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Daily Living and Rehabilitation of Patients with Metastatic
Breast Carcinoma of the Pelvis and Femur

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Table of Contents

| | |
|--|-----------|
| Cover..... | 1 |
| SF 298..... | 2 |
| Table of Contents..... | 3 |
| Introduction..... | 4 |
| Body..... | 4 |
| Key Research Accomplishments..... | 6 |
| Reportable Outcomes..... | 6 |
| Conclusions..... | 6 |
| References..... | 7 |
| Figures..... | 9 |
| Appendix 1..... | 13 |

INTRODUCTION

The subject of the project is to develop models to predict the pathological fracture risk in activities in daily living life (ADL), nursing care, and rehabilitation in breast cancer patients with metastatic lesion in the pelvis and proximal femur. The purpose of the project is to develop a computer model of the pelvis and proximal femur which can be used to predict the pathologic fracture risk and study the effects of pelvic and proximal femoral metastatic bone lesions on the care and management of breast cancer patients. The scope of the research is to include the construction of graphical and quantitative models of the pelvis and proximal femur on computer workstations including Finite Element Method (FEM) and Discrete Element Method (DEM) to study the stress/strain in the pelvis and proximal femur and pressure distribution of the hip joint in the patient with metastatic bone lesions of the breast cancer in the pelvis and proximal femur with interactive capability.

BODY

This section describes the research accomplishment associated with each Task outlined in the approved Statement of Work (written in italic letters).

Technical Objectives 1: Computer model construction

Task 1: Months 1-6: Establishment of database for location, size, and distribution of metastatic breast cancer to pelvic and femoral regions.

Completed as described in the first annual report.

Task 2: Months 6-18: Development of Multi-Discrete Element Model (DEM) and Finite Element Model (FEM) of pelvic and femoral regions based on the database established in Task 1.

An interactive DEM model of the hip joint has been developed as described in the first annual report. As described in Task 6, the loading conditions during the activities daily living (ADL) are now incorporated in the model. A simplified FEM model of the long bone with a bone defect was created (Fig. 1). The results of the FEM was validated by actual mechanical testing (see Task 4). The pelvic FEM model is being developed.

Task 3: Months 6-12: Mechanical testing using cadaveric specimens of pelvic region with and without bone defects in the pelvis.

See Task 4.

Task 4: Months 13-18: Mechanical testing using cadaveric specimens of pelvic region with and without bone defects in the proximal femur.

Both the Dynamic Joint Simulator and MTS BIONIX 858 testing machine are now functioning well at Ross facility. An Acoustic Emission (AE) technique was applied to monitor the microfracture of the cadaveric bone specimen with bone defect during mechanical testing

(Figs. 2 and 3). Mechanical testing was performed on the long bone shaft to validate the results of the FEM model (see Task 2). Fractures were initiated from the corners of the defect under torsional testing where the highest stress/strain was calculated in the FEM model (Fig. 1).

In the original protocol of mechanical testing, we planned to apply loading to the central part of the hip joint. However, the highest pressure in the hip joint was estimated at the posterior part of the hip joint during some of activities in daily living as described in Task 6, and this area appears to be critical for pathologic fractures. Therefore, it is important to identify to the most critical loading condition and apply this loading condition for mechanical testing of the pelvis and proximal femur with bone defect. Detailed loading conditions will be estimated by Tasks 6 and 7; for this reason, Tasks 3 and 4 will be continued in year 3 parallel to Tasks 6 and 7.

Technical Objectives 2: Establishment of the model to predict fracture risk in activities in daily living life, nursing care, and rehabilitation

Task 5: Months 19-22 Acquisition of kinematic and force data in Activity in Daily Living (ADL), rehabilitation program, and nursing care.

See Task 6.

Task 6: Months 23-28: Analysis of the loading conditions during the activities studied in Task 5.

The loading conditions during activities in daily living, including walking, climbing stairs, and chair-up, were obtained and incorporated in the DEM analysis. The contact pressure distribution during the activities has been analyzed on the hip joint without the defect. The preliminary results of the analysis is presented in Fig. 4 (walking), Fig. 5 (climbing stairs), and Fig. 6 (chair-up).

The following tasks will be performed after Month 23 as planned.

Task 7: Months 29-33: Analysis of the stability and stress/strain distribution in the metastatic pelvis and proximal femur under the loading conditions predicted in Task 6.

Task 8: Months 34-36: Preparation of publications.

In addition to the approved Tasks, we have established a virtual interactive model of external fixation of a long bone. This simulation and visualization model could be expanded to simulate fixation of the pelvic fracture. This study has been presented at ASME-BED (American Society of Mechanical Engineering, Bioengineering Division) 2001 Bioengineering Conference, Snowbird, Utah, June 27-July 1, 2001 (**Appendix 1**).

KEY RESEARCH ACCOMPLISHMENT

- The basis of the biomechanical model using FEM to estimate the stress/strain distribution around the bone defect has been developed and validated by the mechanical testing.
- Pressure distribution of the hip joint during walking, climbing stairs, and chair-up was estimated by DEM analysis.

REPORTABLE OUTCOMES

- Manuscripts, abstracts, presentations;

Abstract: Kim YH, Elias JJ, Inoue N, Chao EYS: Study on external fixator adjustability using simulation and visualization model. Proceeding of 2001 Bioengineering Conference, American Society of Mechanical Engineering (ASME), BED-Vol. 50: 237-238, 2001.

Presentation Inoue N, Chao EYS. Computer-aided preoperative planning for acetabular osteotomy and total hip arthroplasty. 1st International Symposium and Workshop on Reconstructive Surgery of the Pelvis. Graz, Austria, April 18-20, 2001.

- Patents and licenses applied for and/or issued;
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- Degrees obtained that are supported by this award;
None
- Development of cell lines, tissue or serum repositories;
None
- Informatics such as databases and animal models, etc;
None
- Funding applied for based on work supported by this award;
None
- Employment or research opportunities applied for and/or received on experiences/training supported by this award.
None

CONCLUSIONS

A simple finite element model with a bone defect was analyzed and the results were validated by mechanical testing. The discrete element model was used to estimate the pressure distribution of the hip joint during Activity in Daily Living (ADL), including walking, climbing stairs, and getting up from a chair.

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(2) Abstracts

Kim YH, Elias JJ, Inoue N, Chao EYS: Study on external fixator adjustability using simulation and visualization model. *Proceeding of 2001 Bioengineering Conference, American Society of Mechanical Engineering (ASME), BED-Vol. 50: 237-238, 2001.*

(3) Book Chapter

Chao EY, Inoue N, Elias JJ, Frassica FJ: Image-based computational biomechanics of the musculoskeletal system. In: Bankman, ed. *Handbook of Medical Imaging*. Chap. 18, PP 285-298. The Netherlands: Academic Press.



Fig. 1 Strain energy density (SED) distribution superimposed over a finite element (FE) model with a defect for a torsional load.

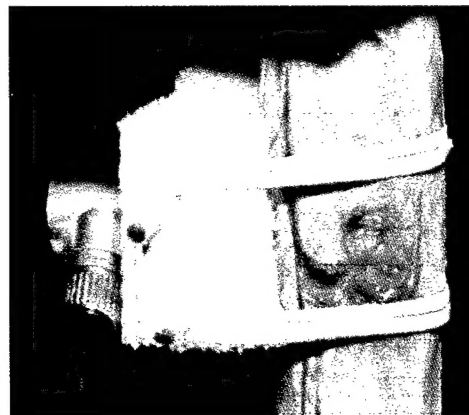


Fig. 2 Torsional failure test of the diaphysis including the defect. An acoustic emission (AE) sensor was attached to the test specimen to detect microfractures during testing.

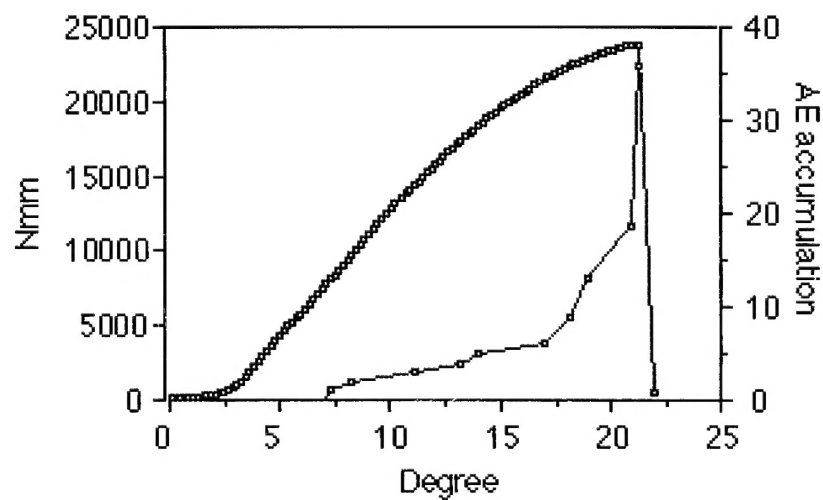
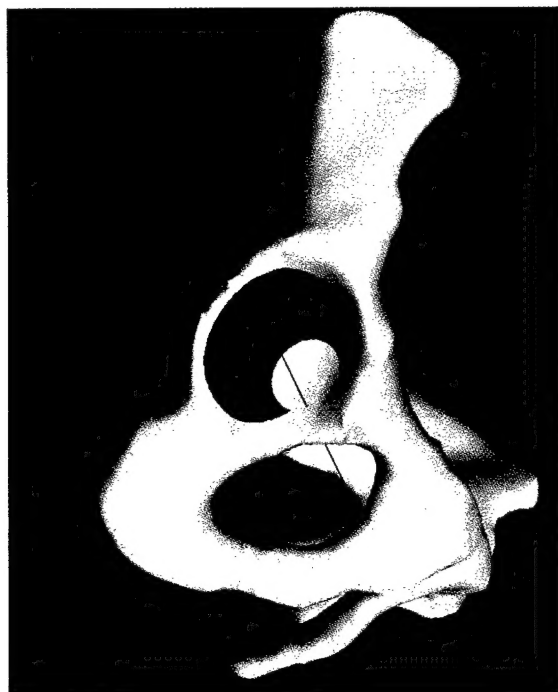
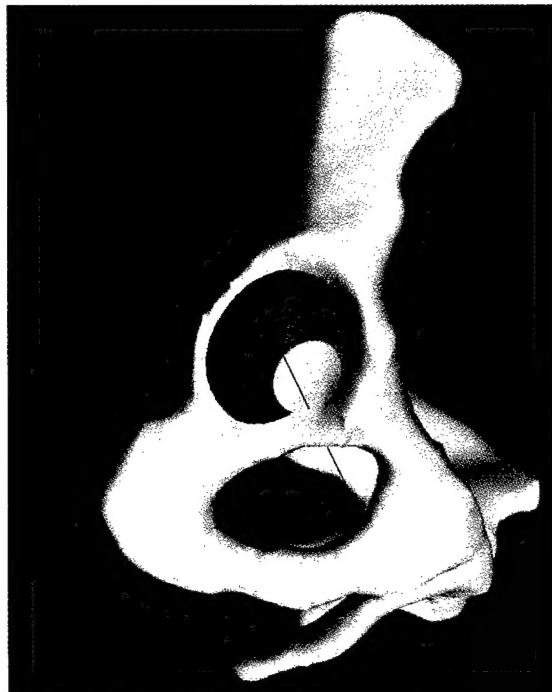


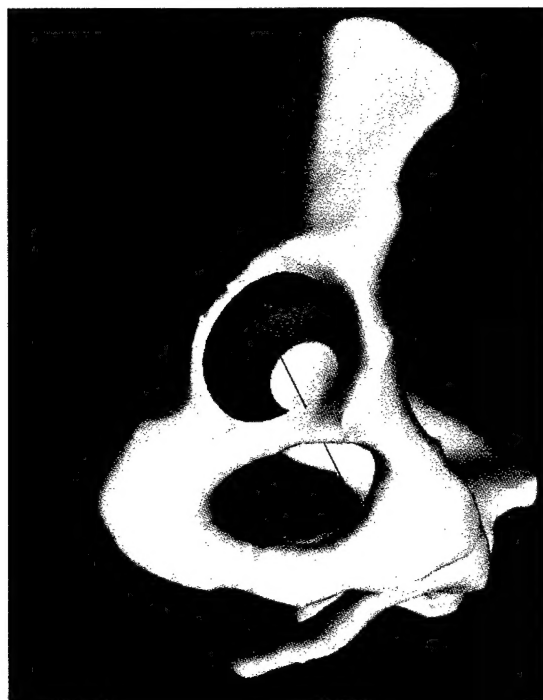
Fig. 3 Torque-torsional angle and AE accumulation-torsional angle curves.



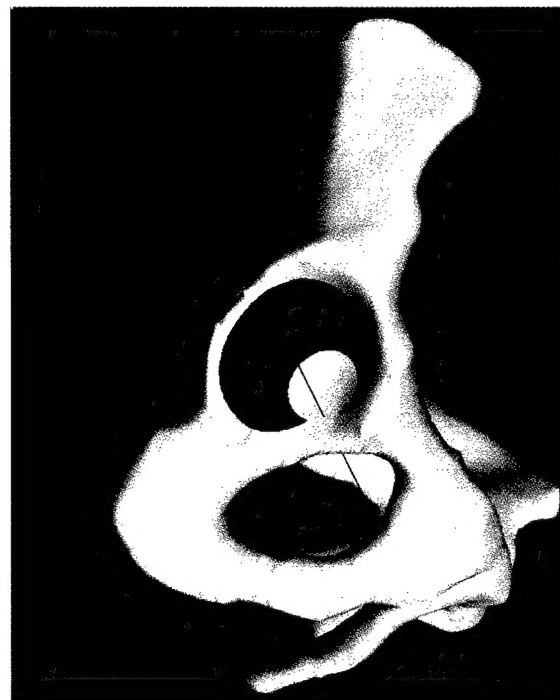
Heel Contact (5%)



Midstance (25%)

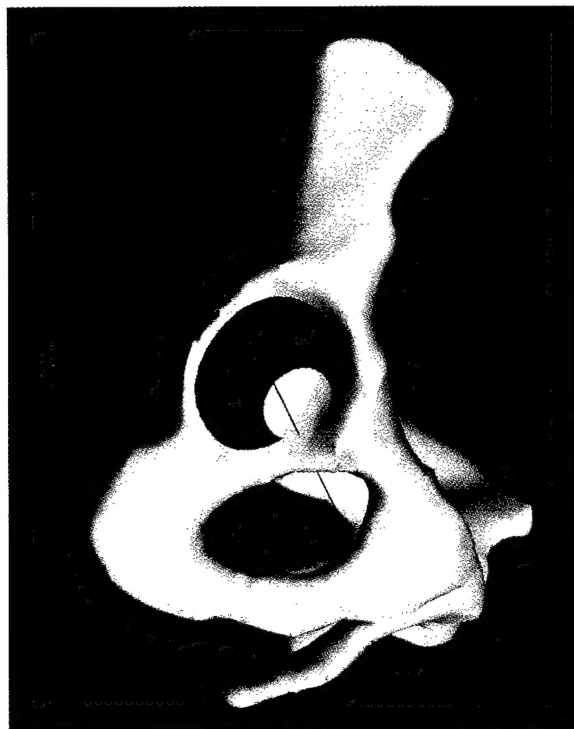


Toe Off (65%)

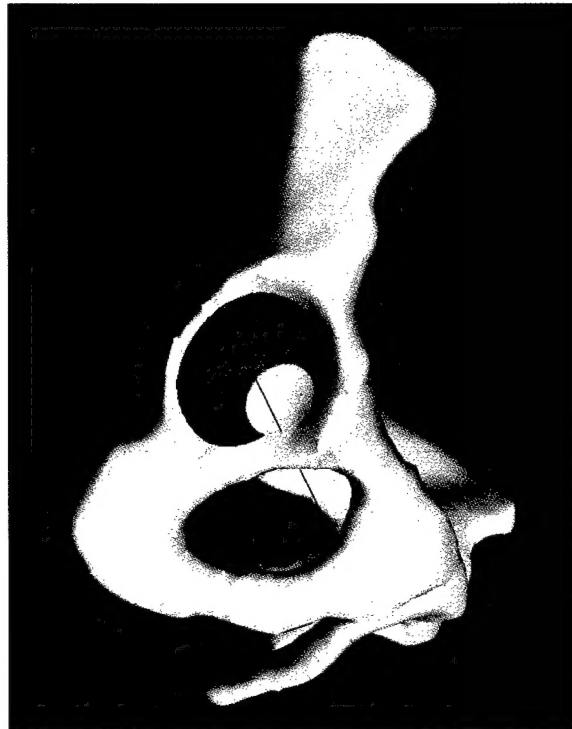


Swing Phase (80%)

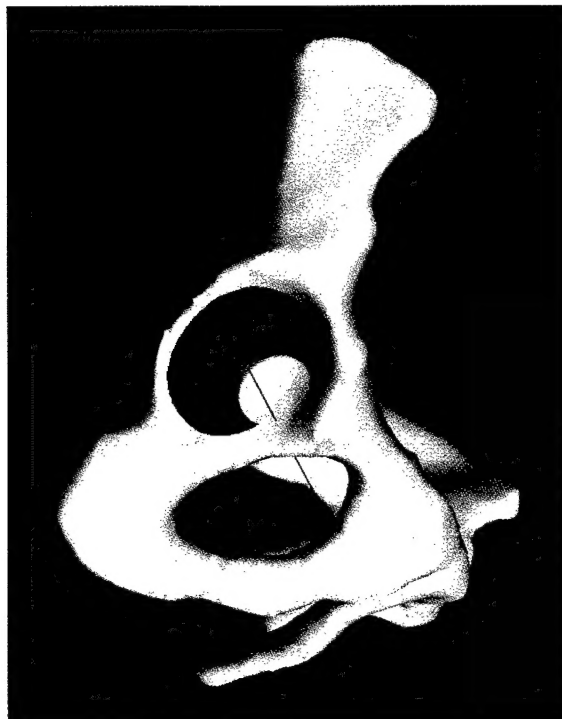
Fig. 4 Showing the contact distribution of the acetabulum during walking of four phases: Heel-Contact, Midstance, Toe-Off, and Swing Phase (% of cycle).



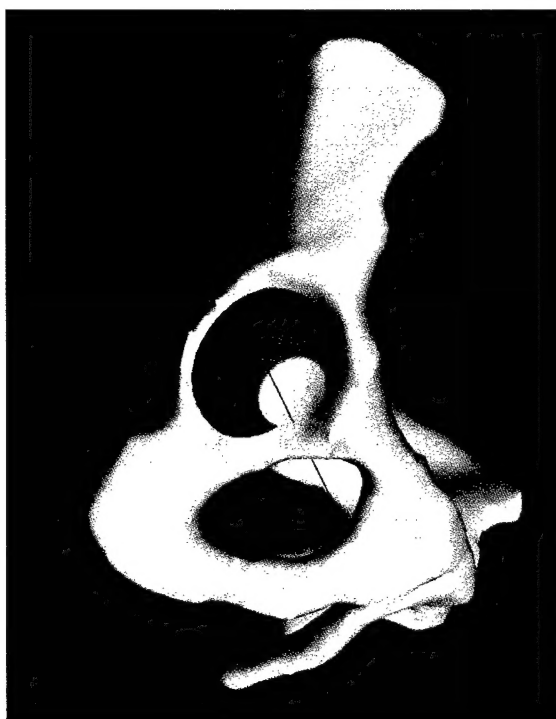
10%



20%

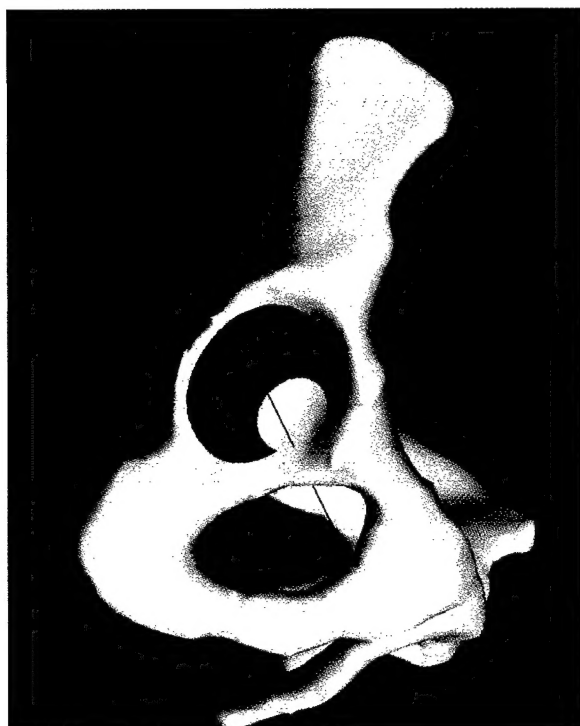


60%

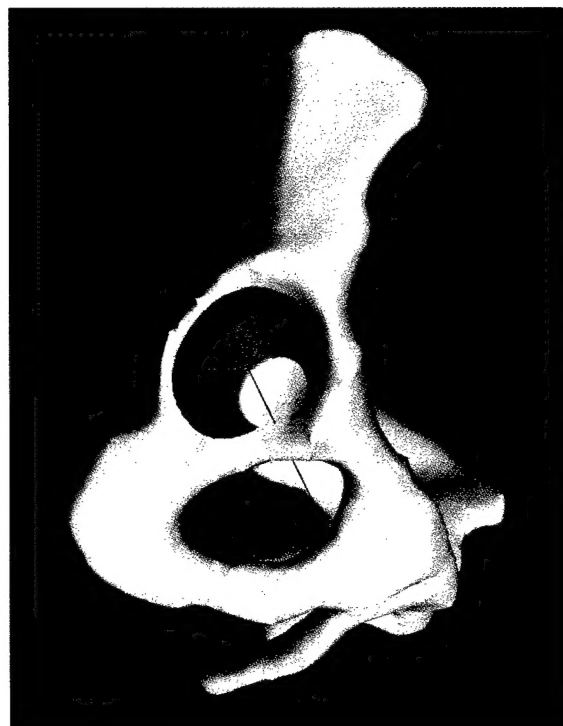


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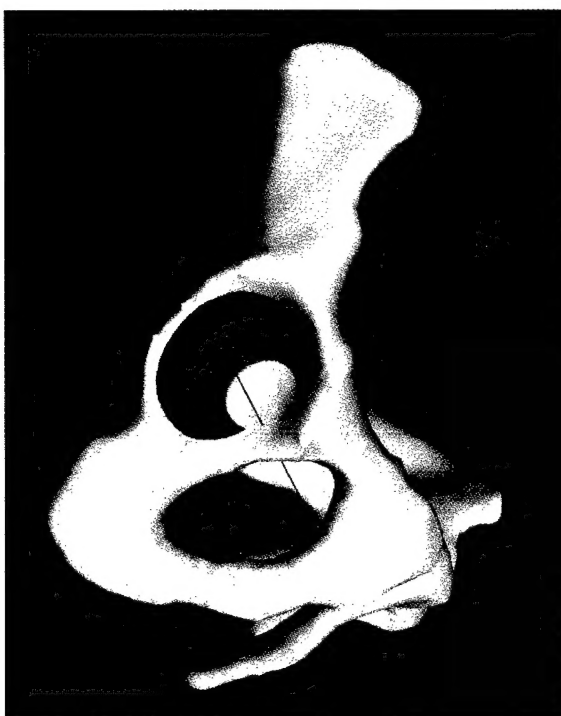
Fig. 5 Showing the contact distribution of the acetabulum during climbing stairs of four phases (% of cycle).



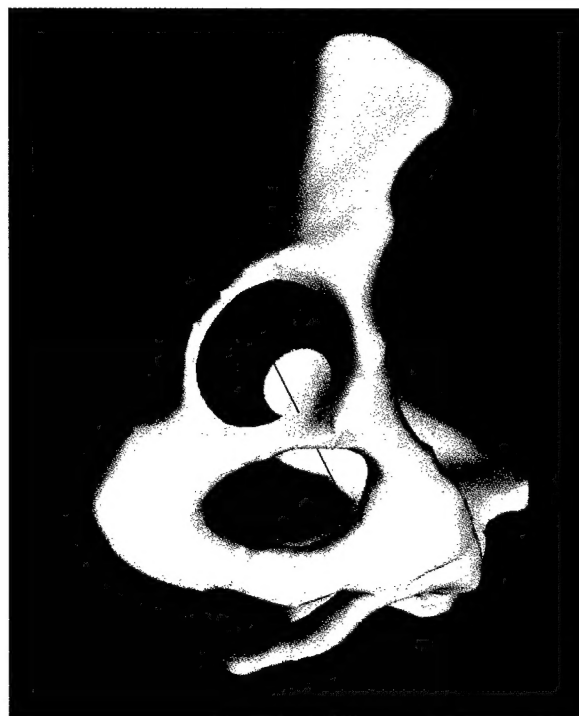
Sitting (5%)



Hip Off (30%)



Inclining (50%)



Standing (80%)

Fig. 6 Showing the contact distribution of the acetabulum during chair up of four phases: Sitting, Hip off, Inclining, Standing (% of cycle).

STUDY ON EXTERNAL FIXATOR ADJUSTABILITY USING SIMULATION AND
VISUALIZATION MODEL

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INTRODUCTION

An external fixator is commonly used to stabilize fractured long bone segments. After inserting the pins and applying the fixator, adjustment of the position of each fragment is often necessary to reduce the fracture site and correct any residual deformities. A primary goal of external fixator design is allowing adequate adjustability for fracture reduction. The combination of an external fixator and a fractured bone can be modeled as a multi-link kinematic mechanism. Using computational techniques, the translations and rotations in the connecting joints of the mechanism necessary to reduce a given fracture can be rapidly quantified. Applying the calculated translations and rotations to a virtual biomechanical model of the fixator and bone system allows 3-D visualization of the adjustments required to reduce a fracture. The objective of this paper is to present a computational and graphical model of an external fixator that can be used to evaluate the fixator design, optimize the design, and guide clinicians using the device.

METHODS

The simulation model was based on an Orthofix, D.A.F external fixator. The model includes the mathematical representation of the motion of each link of the fixator and the bone segments as a kinematic linkage connected by various joints (Fig. 1). The distal bone fragment (D) was fixed to the origin of the global coordinate system (G), with the primary axes of the bone aligned with the global axes. The position of the proximal bone (P) within the global coordinate system was described using a 4×4 homogeneous transformation matrix, ${}^G T_P$, which can be determined radiographically using established bone landmarks [1, 2]. This matrix is equivalent to the transformation of each link of the fixator from the proximal bone segment to the distal bone within the global coordinate system, as:

$${}^G T_P = {}^G T_D \cdot {}^D T_1 \cdot {}^1 T_2 \cdot {}^2 T_3 \cdot {}^3 T_4 \cdot {}^4 T_P, \quad (1)$$

where ${}^G T_D$ is a 4×4 unit matrix representing the position of the distal bone in the global coordinate system, and the other transformations represent transformation of each segment of the linkage (Fig. 1). ${}^D T_1$

and ${}^4 T_P$ represent rigid body translations in the directions of the pins. For the current model, the pin length was set to 40 mm. In addition, ${}^2 T_3$ represents a rigid body translation along the length of the telescoping slider mechanism within the body of the fixator. Also, ${}^3 T_4$ and ${}^1 T_2$ are pure rotations at the ball and socket joints. The rotation sequence of the ball and socket joints for α_i , β_i , and γ_i , $i=1$ and 2 , is Rot(x)-Rot(y')-Rot(z''), as a standard Briant angle convention[3]. The maximum allowable length of the translation for the telescoping slider and the rotation angle for the ball and socket joints are 40 mm and 20° , respectively. The initial malalignment is expressed within ${}^G T_P$ as rotations around the axes of the fragment and a translation along the long axis of the bone. To realign the bones, the resulting system of equations is solved for the rotation or translation at each joint. For this model the length of the bone from the fracture site to the closest pin on both the proximal and distal segments was set to 80 mm.

After substituting in the known values in the transformation matrices, ${}^G T_D$, ${}^D T_1$ and ${}^4 T_P$, Eq. 1 can be reduced as follows:

$${}^1 T_4 = {}^1 T_2 \cdot {}^2 T_3 \cdot {}^3 T_4, \quad (2)$$

where

$${}^1 T_4(1,4) = \cos(\beta_1) * t_2,$$

$${}^1 T_4(2,4) = -\sin(\alpha_1) * \cos(\beta_1) * t_2,$$

$${}^1 T_4(3,4) = \cos(\alpha_1) * \cos(\beta_1) * t_2,$$

where α_1 , β_1 and t_2 are two angle parameters in ${}^1 T_2$ and the translation parameter in ${}^2 T_3$, respectively (Fig. 1). After solving for t_2 , α_1 and β_1 using the three equations in Eq. (2) and calculating γ_1 using the projection method, α_2 , β_2 , and γ_2 are solved using three other equations in Eq. (2). Two clinical cases, a rotational malalignment combined with a fracture gap between two bone segments and a varus angulation, were investigated to validate the model and the analysis involved.

RESULTS

In the first case, a rotational malalignment combined with a fracture gap between two bone segments is considered. The initial gap and the rotational malalignment were set to 6 mm and 30° , respectively. One possible solution is given in Fig. 2 for both the proximal and the distal ball joints as $(\alpha_1, \beta_1, \gamma_1) = (5^\circ, -18^\circ, 16^\circ)$, $(\alpha_2, \beta_2, \gamma_2) = (-10^\circ, -16^\circ, 16^\circ)$ and no extension in the slider (Fig. 2). In the second case, 10° varus angular deformity correction is simulated and visualized (Fig. 3). The candidate solution to correct the deformity was estimated as $(\alpha_1, \beta_1, \gamma_1) = (5^\circ, 0^\circ, 0^\circ)$, $(\alpha_2, \beta_2, \gamma_2) = (5^\circ, 0^\circ, 0^\circ)$ and $t_2 = 12.3$ mm. The graphical models clearly depict where the adjustments were performed to reduce the bone segment malalignment perfectly. It is important to realize that the component adjustments are sequence dependent according to the order of the transformation matrices multiplication assigned. The order of adjustment can be changed which may produce a different solution, thus making the adjustability of a fixator nonunique.

DISCUSSION

External fixators are often applied to polytrauma cases under an environment in which adequate imaging equipment may not be available to provide acceptable fracture segment reduction. Therefore, subsequent adjustment of the fracture ends is often required for a more complex device. Unfortunately, this issue of design requirement rarely has been investigated and this study provides a systematic approach to this problem. Using the kinematic linkage model, the rotational malalignment combined with a fracture gap displacement could be modeled and analyzed. Although the adjustment solution may not be unique, the model quantified the order and the magnitude of translation and rotation for each joint of the fixator to align the bone ends. These calculated adjustments could reduce the fracture malalignment and angulation, but they may not be feasible due to bone fragments collision or to produce excessive soft tissue stretching during reduction. Visualization of the sequential adjustments used to reduce the fracture gap based on alternative solutions should help to determine the clinical feasibility of the reduction process. Further development of the model should include constraints to avoid collision or excessive soft tissue extension. These graphical model and analysis simulation are powerful tools to evaluate and validate the device performance.

The simulation model allows the kinematic analysis results to be visualized. The bone and pin length, and the initial deformity are input to the model. This model and analysis technique can be used to illustrate the necessary fixator component adjustment to correct the malalignment, making the system a valuable preoperative planning tool and a teaching aid. This simulation technology is now being extended to study the strength and stiffness of the fixator under physiological loading. Stress/strain analyses can also be incorporated in the future to reduce the need for mechanical testing of these devices. By incorporating the translational or rotational limit of each joint, the model can also be used to determine the limits of adjustability inherent to a given fixator design.

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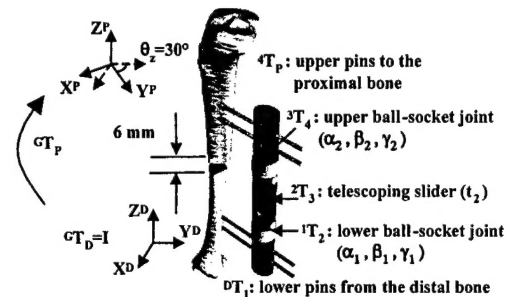


Figure 1. A bone and fixator system with rotational malalignment combined with a fracture gap.

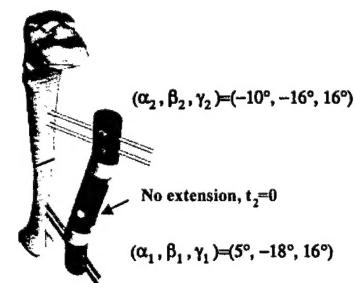


Figure 2. A bone and fixator system after correction.

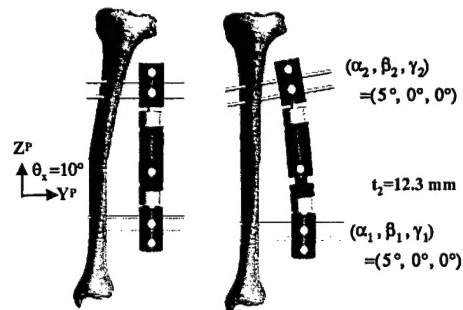


Figure 3. A bone and fixator system in a 10° varus angular deformity correction.